AN APPLICATION OF MICROWAVE HYPERTHERMIA FOR LUNG TREATMENT

Nour El Din Ismail[1], Ahmed M. Abdel Razak[2]

[1]Electrical Eng. Dept., Faculty of Engineering, Alexandria University, Alexandria, Egypt.
[2]Instrument Eng., WEPCO, Alexandria, Egypt
e-mail: ahmed.elshab@gmail.com

***Abstract*- Hyperthermia, particularly when used in conjunction with ionizing radiation, is gaining increasing attention as a cancer treatment modality. If the tumor temperature can be maintained at elevated levels for a specific period of time (to achieve a specified thermal dose), tumor cell killing rate can be increased. Temperature outside of the tumor volume should be kept low enough to avoid damage to normal tissue. Malignant cells are reliably more sensitive to heat than normal cells. Raising the temperature of the tumor is one way to destroy cancer cells.**

Hyperthermia is usually classified as local, regional or whole body [1].

The aim of this paper is to increase the temperature value at a location inside human lung. The temperature distributions around lung are also obtained.

I. INTRODUCTION

The objective of hyperthermia treatment is to raise the temperature at the tumor volume to the range 42°C–45°C for a sufficient period of time in order to achieve cell death or render the cells more sensitive to ionizing radiation and chemical toxins.

The persistent challenges with hyperthermia treatment are preferentially heating the cancerous tissue while maintaining the surrounding healthy tissue at temperatures well below 42°C (i.e., avoiding the introduction of auxiliary foci, or "hot-spots") and noninvasive monitoring the thermal dose delivered to the cancerous tissue [2].

The antenna type used in this paper is a λ/2 dipole. The number of λ/2 dipole antennas are specified according to the operating frequencies (433, 915MHz and 2.45Ghz). A separation distance between two dipoles is λ/2. The dipoles are placed on the suggested model as shown in figure 1.

The dipole length are specified as the following, at frequency f = 433MHz the dipole length L = 34.6cm, at f = 915MHz, L = 16.4cm and at f = 2.45GHz, L = 6.1cm.

Plane wave fields, polarized in the E orientation, were utilized to irradiate the models at exposure frequencies [3].

The electric field radiated by a λ/2 dipole is given as:

$$E_n = E_0 \, . e^{\, i(\omega t \, - \, \beta z \, * \, \theta n)} \tag{1}$$

Where:

E_0 : the amplitude of electric field.

ω : the angular frequency.

β : the phase shift.

θ_n : the phase of the nth dipoles.

The Finite Difference Time Domain (FDTD) technique was used to calculate the specific absorption rate (SAR) inside the biological medium suggested.
SAR is defined as:

$$SAR = \frac{\sigma |E|^2}{2\rho} \tag{2}$$

Where:

|E|: the field inside the medium determined numerically.
ρ : density of tissue (Kg/m^3).
σ : conductivity of tissue (Ω/m).

The temperature distribution at any point inside the medium is obtained by solving the Bioheat Transfer Equation (BHTE).

In section II the human chest model description of different layers and the mathematical tools which used to determine the SAR from the electric field are presented and then the temperature distribution is determined using BHTE. The dielectric constants of each layer at different frequencies are mentioned.

The results are discussed in section III at different frequencies.

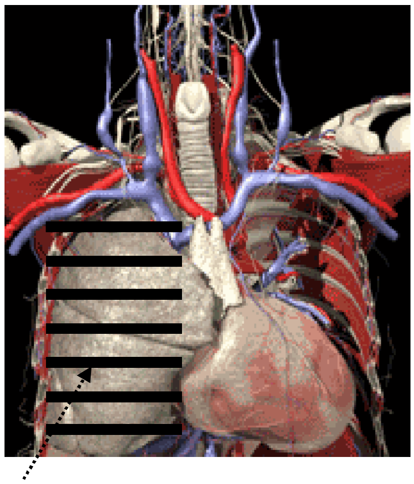
Based on the above results, the conclusion is mentioned in section IV.

II. MODELING AND MATHEMATICAL TOOLS

Chest area is a diverse area of the body. This area contains a large amount of bone, together with some skin, fat, muscle and lung tissues (which contains tumors to be treated without surgical operations) as in figure 1.

Our model is not round as infinite cylinder, but more ellipsoidal and is finite, not in regard to some researches.

Our model is an elliptical shape of biological tissues surrounded by a layer of water bolus, the model is composed of skin, fat, ribs, lung tissue and air as shown:



λ/2 dipole

Fig. 1.a Human Chest

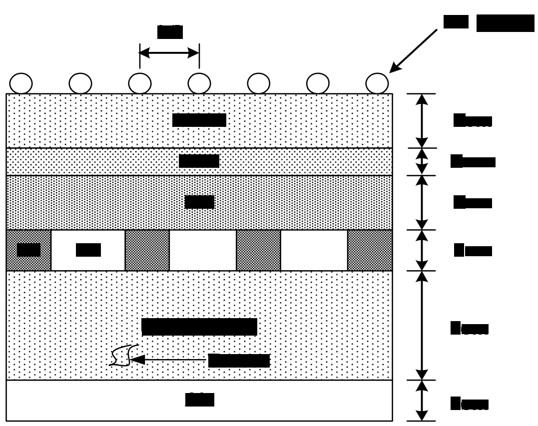


Fig. 1.b Chest Model

The dielectric constants of skin, fat, bone and lung tissues at different operating frequencies are given in table 1. In table 2 the density and the specific heat of skin, fat, bone and lung tissues also are given.

TABLE I

TISSUE TYPES AND THEIR DIELECTRIC CONSTANTS AT 433MHz [4], 915MHz [5] AND 2.45GHz [6]

FREQ.	433 MHz		915 MHz		2.45 GHz	
MEDIA	ε_r	σ (S/m)	ε_r	σ (S/m)	ε_r	σ (S/m)
SKIN	**17.4**	**0.42**	**16.9**	**0.38**	**14.2**	**0.29**
FAT	**5.7**	**0.07**	**5.2**	**0.078**	**4.3**	**0.082**
BONE	**5.7**	**0.07**	**5.2**	**0.078**	**4.3**	**0.082**
LUNG	**35**	**1.1**	**37**	**0.9**	**38.1**	**0.88**

TABLE II

TISSUE TYPES AND THEIR DENSITY AND SPECIFIC HEAT [7]

MEDIA	ρ (Kg/m^3) (TISSUE DENSITY)	*C* (J/(Kg.°C)) (SPECIFIC HEAT)
DI Water	**1000**	**4186**
SKIN	**1100**	**3500**
FAT	**1780**	**1304.832**
BONE	**1785**	**1302.36**
LUNG	**1080**	**3502.74**

The FDTD is introduced and applied to the wave equation, leading to introductory discussions of numerical dispersion, numerical phase velocity and numerical stability, expanding the solution of Maxwell's equations in two and three dimensions.

Once the program determines SAR, BHTE can be solved to get the temperature distributions into the lung tissues.
The BHTE is given as, [8]:

$$\rho c\frac{\partial T}{\partial t} = K\nabla^2 T - W_b C_b (T - T_a) + \rho.SAR \qquad (3)$$

Where:

ρ : density of tissue (Kg/m^3).
c : heat capacity (specific heat) (J/(Kg.°C)).
T : temperature of region (°C).
t : time (sec).
K : thermal conductivity of tissue (W/(m. °C)).
W_b : blood perfusion rate (Kg/(m^3.sec)).
C_b : blood perfusion specific heat (J/(Kg.°C)).
T_a : arterial blood temperature (°C).
∇ : spatial gradient operator.

The BHTE takes into account the exchanges mechanisms such as blood flow rate, heat conduction and metabolic heat generation.

III. RESULTS AND DISCUSSIONS

The following figures show the temperature distribution into the lung tissues. The mesh segment area used in our program is Δx ∗ Δx where Δx = 3mm. According to the stability condition Δt is selected to be 5psec while the time steps = 750 Δt.

The point at which the temperature will be maximum inside the lung tissue is chosen to be (160mm, 140mm).

Figure 2 shows the temperature distribution for the operating frequency 433MHz. It is observed that there are

some points of high temperature values around lung tissues. Figure 3 represents the temperature distribution at frequency 915MHz.

The same observation about many points of high values of temperature is found. When the operating frequency is changed to 2.45GHz, the temperature distribution has less points of high temperature around the lung volume. This observation can be noticed in figure 4.

The bolus used has a conductivity of 0.004S/m and the permitivity of 70 and a thickness is 4cm.

The 3D graph of temperature distribution at operating frequency 2.45GHz is represented in figure 5.

The dipole phases during the calculations are specified as the following:

For 433MHz, there are three dipoles with length 34.6cm and a separation distance between two dipoles as mentioned before. The phases of the three dipoles are (0º, 36º, -38º), also for 915MHz the separation distance is 16.4cm, so that the number of used dipoles is four dipoles with phases (-38º, -24º, 26º, 37.5º).

The last result is at 2.45GHz, the mission used seven dipoles at phases (-28º, -20, -12º, 2º, 14º, 22º, 34º).

It is observed that there is a high temperature occurred in the bolus layer. It means that the bolus layer is essential for cooling the model used.

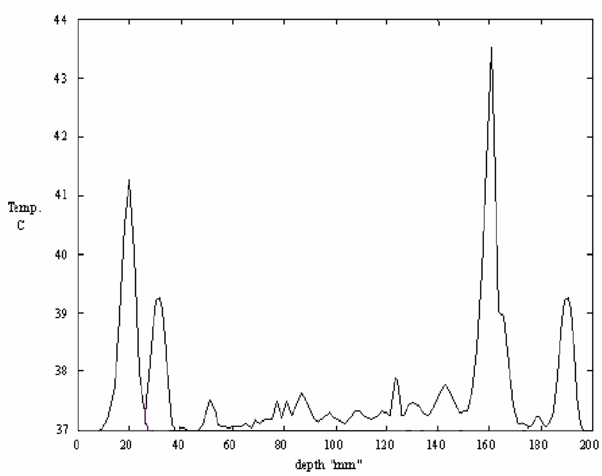


Fig. 2 Temperature distribution at 433MHz for lung tissues.

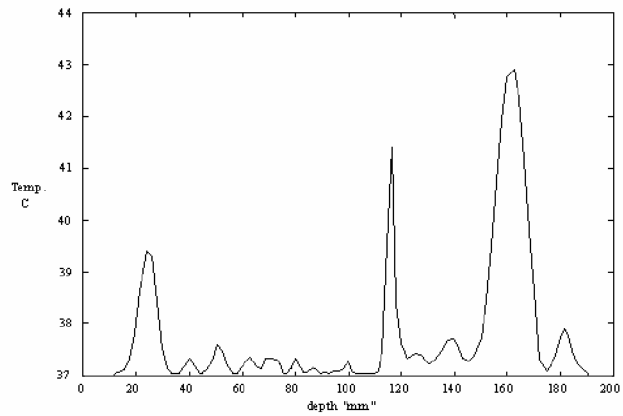


Fig. 3 Temperature distribution at 915MHz for lung tissues.

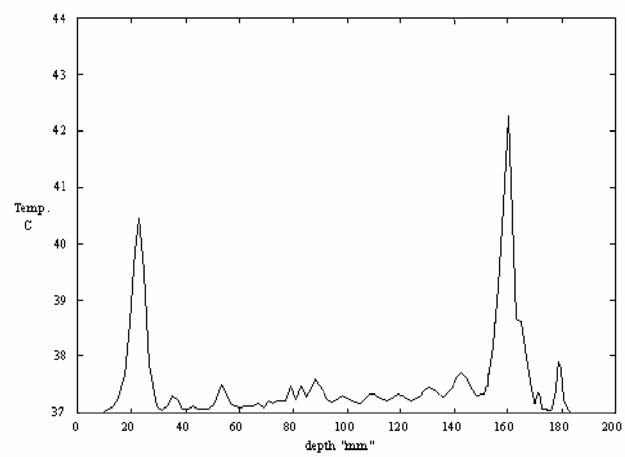


Fig. 4 Temperature distribution at 2.45GHz for lung tissues.

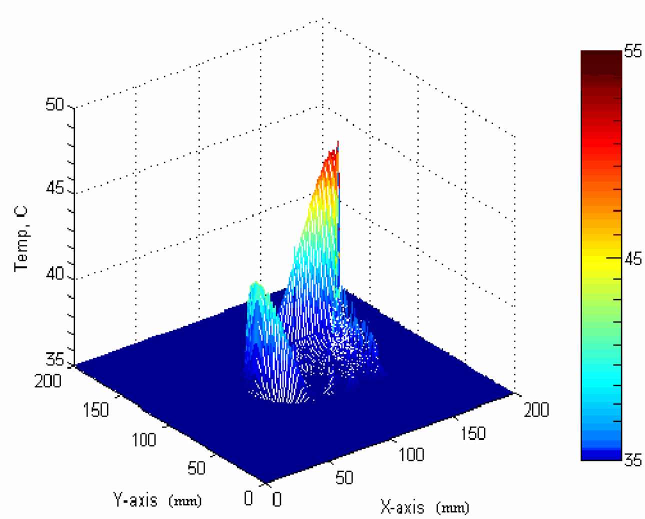


Fig. 5 3D Temperature distribution at 2.45GHz for lung tissues.

IV. Conclusion

The persistent challenges with hyperthermia treatment are preferentially heating the cancerous tissue while maintaining the surrounding healthy tissue at temperatures well below 42°C.

It is observed that the frequency 2.45GHz is the best selected which results high temperature into the target without hot spots maintaining the surrounding healthy tissues at low temperature, skin is maintained at low temperature using bolus layer.

Finally, it is found that microwave hyperthermia can increase the temperature values of certain locations inside lung tissues. This process can be used to kill malignant tissues inside the lung region without surgery as a noninvasive treatment technique.

References

[1] J. de Bree and J. Lagendijk "3-D SAR Model for Current Source Interstitial Hyperthermia", Vol. 43, No. 10, Oct. 1996.

[2] D. Tompkins, R. Vanderby and W. Beckman " Effect of Interseed Spacing, Tissue Perfusion, Thermoseed Temperatures in Ferromagnetic Hyperthermia" Vol. 41, No. 10, Oct. 1994.

[3] GeoCities HYPERLINK 5th Edition "http://www.esho.info/patients/faqs" M. J. Piket, A. Taflove, W. C. Lin and D. S. Katz, 2002.

[4] C. Wang "Numerical simulation of annular phased array for anatomically based models using the FDTD method" *IEEE Trans. Biomed. Eng.*, Vol. 37, pp. 118-125, January 1989.

[5] E. Lee and T. Wilsey "Body Conformable 915MHz Microstrip Array Applicators for Large Area Hyperthermia" Vol. 39, No. 5 May 1992.DENNIS M.

[6] SULLIVAN, GANDHI "Use Of The Finite Difference Time Domain Method In Calculating EM Absorption In Human Tissues" *IEEE Trans. Biomed. Eng.*, Vol. 34, No. 2. Feb. 1987.

[7] R. M. Arthur, W. L. Straube, J. D. Starman, and E. G. Moros, "Noninvasive Temperature Estimation Based on the Energy of the Backscattered US" *Med. Phys.*, Vol. 30, No. 3, pp. 1021–1029, June 2003.

[8] K. Nikita, N. Maratos and Nikolaos papeUzunoglu "Optimal Steady-State Temperature Distribution for a Phased Array Hyperthermia System" Vol. 40, No. 12, Dec. 1993.